Analysis of Viscoelastic and Poroelastic Behavior in MR elastography

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INTRODUCTION Elastography techniques are based on a mechanical model of tissue, for example, equation 1 represents an isotropic, viscoelastic (VE) solid. A recent development in MRE is the introduction of poroelasticity (PE, Eq. 2), which models tissue as a porous elastic matrix with an infiltrating pore fluid, and is regarded as an appropriate model for the biphasic environment of brain tissue[1], and in the physiology literature more generally for all tissues[2]. We are pursuing 2 different sources of motion for brain MRE: Standard external actuation (EA)[3], where the motion is provided by an actuator at 50-125Hz, and intrinsic actuation, (IA)[4], which uses the natural pulsatility of the brain as the motion source (at ~1Hz). This work examines the suitability of VE and PE assumptions for each case.

$$\nabla \cdot \mu(\nabla\vec{u} + \nabla\vec{u}^T) + \nabla(\lambda\nabla\cdot\vec{u}) = -\rho\omega^2\vec{u}$$

Eq 1: Non-homogeneous form of Navier's Equation, governing motion in a solid with spatially varying viscoelastic properties, where $\vec{u}$ = 3D displacement vector, μ = complex valued shear modulus, λ = compressional modulus, and ρ = density.

$$\nabla \cdot \mu(\nabla\vec{u} + \nabla\vec{u}^T) + \nabla(\lambda\nabla\cdot\vec{u}) - (1-\beta)\nabla P = \left(\rho - \beta\rho_f\right)\ddot{\vec{u}}$$
$$\omega^2(1-\beta)\nabla\cdot\vec{u} + \beta\nabla^2 P = 0$$

Eq 2: Governing Equations of poroelasticity. In addition to terms in Eq 1, P is the fluid pressure, ρ_f is the fluid density, and β is a parameter combining a range of poroelastic constants such as porosity (ϕ) and hydraulic conductivity (κ).

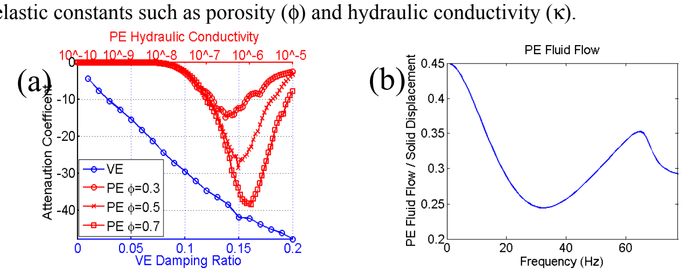

Figure 1a: Exponential attenuation coefficient for free vibration of 3D PE and VE beams. PE values are plotted as functions of κ for 3 different ϕ values.

Figure 1b: Relative size of PE fluid flow for a constrained system, plotted as a function of frequency. The peak at 65Hz corresponds to the first resonant mode.

DATA AND METHODS Numerical simulations are used to show that VE and PE models can produce similar results during free vibration at the frequencies associated with EA MRE, and that PE effects become more significant with the physical constraints and low frequencies of IA. These results are used to explain differences in the performance of VE and PE MRE reconstructions of healthy volunteers for 50Hz EA and 1Hz IA. Reconstructions of 5 different subjects are analyzed for each actuation technique.

RESULTS Fig 1a. is a plot of the attenuation coefficient for free vibration of 3D VE and PE beams, and Fig. 1b shows the relative size of the PE fluid flow for a physically constrained system such as the cranium. Figs. 2 and 3 compare VE and PE MRE reconstructions of representative 50Hz EA and 1Hz IA brain datasets, respectively.

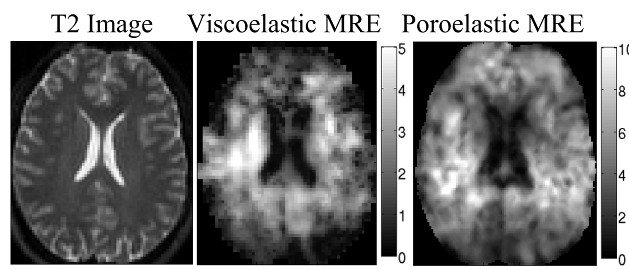

Figure 2: Externally actuated 50Hz MRE reconstructions of healthy human brain. Both VE and PE reconstructions show anatomic structures, PE stiffness estimates are generally higher than other published stiffness values, possibly due to the required assumptions of hydraulic conductivity and porosity values.

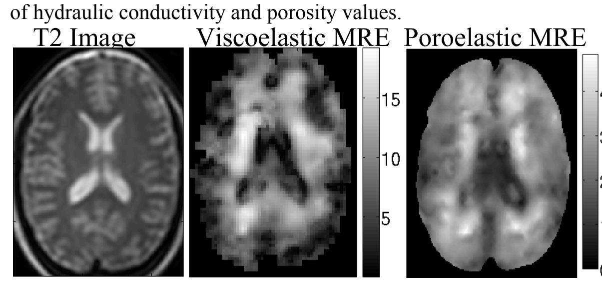

Figure 3: Intrinsically actuated (1Hz) MRE reconstructions of healthy human brain. PE results are consistent between subjects and values fall within the range commonly reported by MRE. VE reconstructions of IA data find some anatomical features, however, they are prone to artifacts and intersubject consistency is poor.

CONCLUSIONS Previous MRE studies have estimated a VE damping ratio of about 0.2[5]. Fig 1a shows the resultant attenuation is on the order predicted by PE when a significant portion of the tissue is mobile fluid (high porosity). EA PE and VE reconstructions both find anatomical structures, with PE overestimating shear modulus (relative to the literature), likely due to incorrectly assumed values of ϕ and κ, causing the model to underestimate the true level of attenuation. Fig 1b shows PE fluid flow in a constrained system is maximized at very low frequencies, and fig. 3. confirms that PE reconstructions are preferable at the low frequency of IA. These results suggest treating brain tissue as VE at higher frequencies may be an adequate approximation to its biphasic behavior[2,6], and accurate use of PE requires values of physiological parameters such as κ and ϕ, which are difficult to obtain (although it may be possible to reconstruct them using PE MRE[7]). However, at low frequencies PE reconstructions produce better results due to more accurate modeling of the redistribution of fluid within the cranial compartment resulting from cyclic changes in fluid and arterial pressure within the parenchyma caused by vascular pulsations during the cardiac cycle.

REFERENCES :

1: Perriñez et al.: *IEEE Trans Biomed Eng* 2009.
2: Aukland and Reed Physiolgical Reviews 1993
3: Weaver et al.: *Med Phys* 2001
4: Paulsen et al.: *Proc. ISMRM* 2011,
5: Sack et al.: *NMR in Biomed.* 2008
6: Kyriacou et al.: Biomechan Model Mechanobiol 2002.
7: Pattison et al: *Proc ISMRM* 2011.